Elsevier Editorial System(tm) for Gait and Posture

Manuscript Draft

Manuscript Number:

Title: Full gait cycle analysis of lower limb and trunk kinematics and muscle activations during walking in participants with and without ankle instability

Article Type: Original Paper

Keywords: 3D kinematics, Electromyography, Chronic ankle instability, Gait

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Yours faithfully,

Lynsey Northeast

DECLARATION OF INTEREST: The authors report no conflicts of interest. The authors alone are responsible for the content and writing of the paper.

A comparison of lower limb kinematics and electromyography during walking between athletes with chronic ankle instability and healthy controls

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Abstract

Introduction: Chronic ankle instability (CAI) has previously been linked to altered lower limb kinematics and muscle activation characteristics during walking, though little research has been performed analysing the full time-series across the stance and swing phases of gait. The aim of this study was to compare trunk and lower limb kinematics and muscle activity between those with chronic ankle instability and healthy controls.

Methods: Kinematics and muscle activity were measured in 36 participants Eighteen (14 males, 4 females) healthy controls (age 22.4 ± 3.6 years, height 177.8 ± 7.6 cm, mass 70.4 ± 11.9 kg, UK shoe size 8.4 ± 1.6), and 18 (13 males, 5 females) participants with chronic ankle instability (age 22.0 ± 2.7 years, height 176.8 ± 7.9 cm, mass 74.1 ± 9.6 kg, UK shoe size 8.1 ± 1.9) during barefoot walking trials, using a combined Helen Hayes and Oxford foot model. Electromyography (EMG) was recorded for the tibialis anterior and gluteus medius. Full curve statistical parametric mapping was performed using independent and paired samples t-tests.

Results: No significant differences were observed in kinematic or EMG variables between or within groups for the duration of the swing phase of gait. A significantly increased forefoot-tibia inversion was seen in the CAI affected limb when compared to the CAI unaffected limb at 4-16% stance ($p = 0.039$). No other significant differences were observed.

Conclusions: Participants with CAI exhibit increased inversion patterns during the stance phase of gait in their affected limb compared to their unaffected limb. This may predispose those with CAI to episodes of giving way and further ankle sprains. Kinematic and EMG analysis of gait using full curve analysis revealed no significant differences between CAI and healthy control groups.

Key Words: 3D kinematics, Electromyography, Chronic ankle instability, Gait

1. Introduction

Lateral ankle sprains are one of the most common musculoskeletal injuries in both general and sporting populations [1, 2]. Following an acute ankle sprain, it is suggested that 32-74% of individuals have residual symptoms such as recurrent sprains, episodes of giving way and/or perceived instability [3]. Chronic ankle instability (CAI) is defined as ‘a history of recurrent ankle sprains and the sensation of giving way’ [4]. Long term, links have been established between the development of osteoarthritis and a history of chronic ankle instability [5]. Developing greater understanding of the biomechanics associated with CAI may aid the development of preventative measures.

Walking is a task of high importance in daily life and is often problematic for people with chronic ankle instability, who complain of a giving way sensation on both uneven and level surfaces [6]. Research suggests that the position of the affected ankle joint at specific time points during the gait cycle may predispose an ankle to injury [7]. This may be associated or caused by ankle joint instability. Furthermore, gait analysis is used to assist with the development of rehabilitation and injury prevention protocols [8], and therefore any changes in gait need to be investigated and where possible, accounted for.

Previous literature investigating CAI during walking has modelled the foot as one rigid segment [9, 10], however, the foot is composed of 26 bones and 20 articulated joints with a number of complex interactions [11]. Rigid segment modelling therefore excludes motion between different segments of the foot providing inadequate information on the biomechanics of the foot [9]. De Ridder et al. [12] appears to be the first study to analyse walking using a multi-segmental foot model. They compared the use of the Ghent Foot Model to a rigid foot model in participants with CAI, copers (suffered from a recent ankle sprain but no symptoms of instability) and control participants. The results of the study lead the authors to conclude that the multi-segmental foot model provided greater details of the intricacies of the foot, showing differences between the segments of the foot when comparing across the groups. Upper body kinematic analysis should be considered when investigating any changes in the lower extremities as there may be a significant relationship with changes observed in proximal segments [13]. To the authors knowledge, no research has combined upper body kinematics with a full lower limb and multi-segmental foot model to address, in combination, the possible proximal and distal differences between groups.

Hip abductor weakness has been associated with acute ankle sprains though it is unclear whether this is a cause or an effect of the sprain [14]. Koldenhoven et al. [15] reported increased gluteus medius activation in the late stance and early swing phase of walking in CAI participants. They suggested that this may be a coping mechanism used to generate a wider base of support or to greater stabilise the lower limb. A decreased tibialis anterior activation was also observed resulting in a plantarflexed position pre-heel strike. This loose-packed position has been found to be unstable [16] suggesting an increased risk of ankle sprains.

Prior research only reports joint angles and muscle activation characteristics at discrete time points during a walk [10, 15], rather than reporting the whole kinematic time-series curves. This may result in focus bias or missing potential significance or trends during other phases of the gait cycle [17]. Statistical parametric mapping (SPM), a concept introduced to biomechanics from brain research [18] enables curve analysis across the whole movement [17] therefore enabling full comparison of the movement. De Ridder et al. [12] used statistical parametric mapping to compare foot kinematics between participants with CAI, copers and controls identifying exact time periods of significance within the stance phase of walking.

The aim of this study was to compare trunk, hip, knee and multi-segmental foot kinematics and muscle activation during the stance and swing phase of walking between participants with CAI and healthy controls.

2. Methods

2.1 Participants

Eighteen (14 males, 4 females) healthy controls (age 22.4 ± 3.6 years, height 177.8 ± 7.6 cm, mass 70.4 ± 11.9 kg, UK shoe size 8.4 ± 1.6), and 18 (13 males, 5 females) participants with chronic ankle instability (age 22.0 ± 2.7 years, height 176.8 ± 7.9 cm, mass 74.1 ± 9.6 kg, UK shoe size 8.1 ± 1.9) participated in this study. Ethical approval was granted by the institutional ethics committee prior to testing. Written informed consent was obtained from participants and a health screen questionnaire was completed prior to participation.

Participants were included in the study if they were aged 18-35 and took part in team sport a minimum of twice a week. Participants were excluded if any of the following applied; acute lower limb injury in the past 3 months, use of prescribed or shop bought orthotics, lower extremity biomechanical abnormality, balance or motion disorders, history of fracture requiring realignment or history of lower extremity surgery in accordance with selection criteria outlined by the International Ankle Consortium (IAC) [3].

Participants were allocated into the control group or the chronic ankle instability group based on results of the Identification of Functional Ankle Instability (IdFAI) questionnaire, where a score of ≥11 indicated ankle instability in accordance with guidelines from the IAC [3]. The affected limb was determined as the self-reported weaker limb in subjects with CAI. Due to the researcher being blinded to the questionnaire outcome, the affected limb could not be identified exclusively as either the dominant or non-dominant limb. Therefore, the affected limb was randomly matched to a control group to adjust for the dominance effect. Limb dominance was determined by asking which leg they would use to kick a ball [16, 19].

2.2 Protocol

Each participant completed a 5-minute warm up on a cycle ergometer (Monark Ergomedic 874E, Sweden) at a steady state of 60 watts. Electromyographic data were recorded bilaterally for the gluteus medius and tibialis anterior using a DataLINK data acquisition system (Biometrics Bluetooth unit W4X8, Biometrics Ltd, Gwent, UK) sampling at 1000Hz with pre-amplified SX230-1000, stainless steel surface electrodes with a 20-450Hz bandwidth, gain x1000, noise < 5 μV, input impedance > 1015 Ω. Participants' skin was prepared for electrode placement by dry-shaving using disposable razors and cleansing with an alcohol wipe. Electrodes were placed in accordance with SENIAM guidelines [20]. The tibialis anterior electrodes were placed at a third of the line between the tip of the fibula and the tip of the medial malleolus. The gluteus medius electrodes were placed half way between the crista iliaca and the trochanter. To allow for comparison between participants, EMG data were normalised to a maximum voluntary isometric contraction (MVIC). For each muscle, three maximal contractions were performed for a 5 second duration with a 1-minute rest between trials. The peak activation for these three trials was then used for subsequent analysis. The gluteus medius MVIC was performed in side lying. The participant was instructed to maximally abduct their hip (positioned in mid-range) into a rigid strap positioned just above the knee [21]. The tibialis anterior MVIC was performed in a seated position and the participant was asked to maximally dorsiflex and invert their foot against a rigid strap [21].

Motion analysis data were recorded using an Owl Digital Real Time 10 camera system (Motion Analysis, Santa Rosa, California) sampling at 200 Hz. The motion analysis system was calibrated as per the system instructions prior to each day of testing. Passive reflective markers were attached to the participant in accordance with the Helen Hayes marker set [22] combined with the Oxford foot model [23]. Markers were attached directly to the skin using double-sided marker stickers. Marker and electrode placement was performed by the same person for all participants.

Participants were instructed to walk at their normal walking speed through the calibrated capture volume. Pace was not controlled, as this was deemed to be unnatural for participants and has been previously shown to have an impact on stride time variability due to increased central nervous system involvement [24]. Participants walked barefoot 3.5 m before data were collected [25] and proceeded for 7 m across the walkway. Walking speed was, however, recorded using the pelvis segment velocity. Each participant performed a familiarisation of the movement until they were comfortable with the movement before recording and three trials were recorded for analysis. A successful trial was deemed as one when all tracking markers were in view of the cameras and where there was no evidence of gait modification. Trials where gait modification occurred were discarded and re-tested.

2.4 Data and Statistical Analysis

Data were inspected using Cortex (Cortex-64 5.3.1.1543, Motion Analysis Corporation, Santa Rosa, California) software before importing into Visual 3d (Visual3D v6 x64, C-motion, Germantown, Maryland). Data were smoothed using a 6 Hz Butterworth filter. As no force plate was used, initial contact was determined using the method proposed by O'Connor et al. [26]. This method creates a new signal by calculating the midpoint between the posterior inferior heel marker and the toe marker (between 2nd and 3rd metatarsal heads). The first derivative was then calculated on the vertical component of the signal. Event markers were created at the minimum value for heel strike and the maximum value for toe off. Electromyographic data were root mean squared by a moving window of 100 ms and normalised to MVIC [27]. Following this a visual inspection of the data identified noise in the signal for two of the participants that warranted their EMG data be removed. In order to keep the pre experimental research design the matched controls assigned to the two participants also had their EMG data removed. Kinematic and EMG data were exported for the stance (heel strike to toe off) and swing (toe off to heel strike) phases into MATLAB R2015a (The Math Works, Natick, Massachusetts) to perform the SPM analysis.

Kinematic data were exported for forefoot-hindfoot angle (FFHFA), forefoot-tibia angle (FFTBA), hindfoot-tibia angle (HFTBA), hip, knee and trunk angles in the sagittal, frontal and transverse planes of motion. So not to eliminate inherent variations in foot morphology data was not normalised against a reference segment [12, 23]. Data were analysed using SPM [18] in MATLAB using the SPM1D open-source package (spm1d.org). Data were tested for normality using a D'Agostino-Pearson's test. A matched control limb was compared to the chronic ankle instability groups affected limb using an independent samples t-test. The unaffected and affected limb of the chronic ankle instability group were then compared using a paired samples t-test (α= 0.05). A matched control limb was also compared to the chronic ankle instability groups unaffected limb using an independent samples t-test (α= 0.05).

3. Results

Independent samples t-tests for full kinematic curve analysis revealed no significant differences ($p > 0.05$) between groups for age, stature, mass, or shoe size. An independent samples t-test reported no significant difference in walking velocity when comparing the control group (1.20 ± 0.15 m/s), and CAI group (1.18 ± 0.09 m/s).

No significant differences were observed in FFHFA, FFTBA, HFTBA, hip, knee, or trunk angles in the sagittal, coronal, or transverse planes of motion, in the stance or swing phase, between the matched control and the CAI groups affected limb. No significant differences were observed in the gluteus medius or tibialis anterior muscle activation in either phase of gait between the matched control and the CAI groups affected limb.

A significant difference was reported between the CAI groups' unaffected and affected limb in the forefoot-tibia angle, where increased inversion was observed in the affected limb at 4-16% of the stance phase (mean difference of 3.07°, p = 0.039, Figure 1.). No other significant differences were reported in the plantarflexion/dorsiflexion or internal/external rotation angles of the forefoot-tibia in the stance or swing phases. Furthermore, no significant differences were noted between FFHFA, HFTBA, hip, knee, or trunk angles or in muscle activation of the tibialis anterior and gluteus medius between the unaffected and affected limbs at any time point. Finally, no significant differences were observed between the CAI groups' unaffected limb and the control groups' limb (matched for dominance) in any of the recorded variables in either the stance or swing phases of movement.

FIGURE 1 AROUND HERE

4. Discussion

The aim of this study was to explore the differences in kinematics and muscle activation patterns between CAI participants' unaffected and affected ankles and to compare the same variables to a matched control group throughout the full gait cycle.

Increased FFTBA inversion was found in the affected limb of the CAI group when compared to its unaffected counterpart at 4-16% stance. This finding is of particular clinical interest as it supports previous hypotheses that participants with CAI may exhibit altered joint position sense and proprioceptive awareness [28]. Increased inversion at ground contact decreases the bony restriction of the foot-ankle complex, thus, when loaded with bodyweight increases inversion torque and the joints susceptibility to injury [28]. The early period of the stance phase is beyond conscious control [7, 10] and thus this increased inversion places the ankle in a position of increased vulnerability at heel strike, potentially predisposing the affected limb to further ankle sprains and episodes of giving way. Whilst not within the remit of this study, the difference in angular displacement associated with CAI may indicate larger difference in more dynamic movements such as cutting, single and double leg landing and running or on when walking on uneven surfaces.

The lack of any significant differences at the hip or knee between groups in the frontal, sagittal or transverse planes of motion in the current study is consistent with the findings of Monaghan et al. [10]. They found no significant differences in hip and knee kinematics between participants with CAI and healthy control participants from 100 ms pre-heel strike to 200 ms post-heel strike. Within the current study, trunk kinematics were measured in all three planes, however, no significant differences were identified between groups throughout the gait cycle. This suggests that no proximal adaptations took place within the CAI group whilst walking.

No significant differences were observed in either tibialis anterior or gluteus maximus muscle activation between groups in either the stance or swing phase of gait. This is contrary to the findings of Hopkins et al. [16] who when reporting discrete peak value data, observed an increase in tibialis anterior activation from 15-30% and 45-70% stance, which they speculated was a motor strategy to maintain a more dorsiflexed, stable position in the affected limb when compared to a dominance matched control limb. Methodological differences exist between the current study and the study by Hopkins et al. [16] as participants walked shod rather than barefoot as in the present study. Decreased muscle activation patterns have previously been observed in barefoot walking when compared to shod walking [29]. Hopkins et al. [16] also examined tibialis anterior activation whilst walking on a treadmill rather than over ground. These differences in methodological approaches may account for the differing results between the two studies. Koldenhoven et al. [15] recorded significantly higher gluteus medius muscle activation in the final 50% of stance and the first 25% of the swing phase, when compared to healthy participants, however, this was again performed shod on a treadmill, so comparisons with the current study are difficult. Previous studies have found different muscle activation patterns and sagittal plane motion with treadmill walking compared to over ground walking [30]. Therefore, the results from this study may prove a more valid representation of the everyday task of over ground walking. It should be noted that due to the different statistical analysis performed in this study compared to that of this previous research how valid comparisons are is not clear, with a lack of research comparing the different methodologies leaving this currently open to speculation [15, 16, 29, 30].

This study analysed kinematic and electromyographic parameters to determine differences in movement patterns and muscle activations, however, future research should identify the impact of CAI on kinetic parameters using full curve analysis to identify if differences exist between groups. Further research should use these analysis methods to examine more dynamic movements such as change of direction, both single and double leg landing and running gait.

5. Conclusion

Participants with CAI exhibit increased inversion patterns during the stance phase of gait in their affected limb when compared to their unaffected limb. This change in movement pattern may predispose those with CAI to episodes of giving way and further ankle sprains. The increased inversion may also be a significant risk factor in more dynamic movements, thus further research should investigate these with use of a multi-segmental foot model. It may also be beneficial to incorporate kinetic variables into this analysis to determine if differences in ground reaction forces and moments are present.

Conflict of interest

None.

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